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Ball et al.

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(54) **MRI SAFE ACTUATOR FOR IMPLANTABLE FLOATING MASS TRANSDUCER**

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(51) **Int. Cl.**
H04R 25/00 (2006.01)

(52) **U.S. Cl.**
CPC ***H04R 25/606*** (2013.01); *H04R 2225/67* (2013.01); *H04R 2460/13* (2013.01)

(58) **Field of Classification Search**
CPC H04R 25/00; H04R 25/40; H04R 25/402; H04R 25/48; H04R 25/55; H04R 25/60; H04R 25/604; H04R 25/606; H04R 25/608; H04R 25/65; H04R 2225/025; H04R 2225/49; H04R 2225/67
USPC 600/25; 607/57
See application file for complete search history.

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Assistant Examiner — Thaddeus Cox
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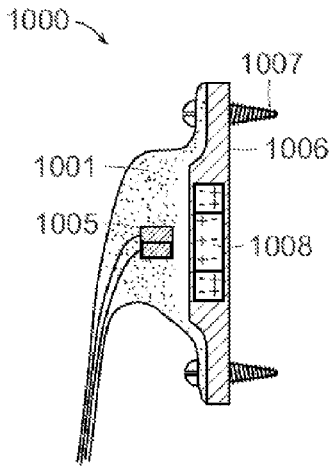
(57) **ABSTRACT**

An implantable hearing prosthesis for a recipient patient is described. An implantable signal transducer includes one or more electromagnetic drive coils for receiving an electrical stimulation signal and a cylindrical transducer magnet arrangement including an inner disk magnet having a first magnetic field direction, and an outer annular magnet surrounding the inner rod magnet and having a second magnetic field direction opposite to the first magnetic field direction. Current flow through the one or more electromagnetic drive coils from the electrical stimulation signal creates a coil magnetic field that interacts with the magnetic fields of the transducer magnet arrangement to create vibration in the transducer magnet which is developed by the signal transducer as a mechanical stimulation signal for audio perception by the patient.

5 Claims, 12 Drawing Sheets



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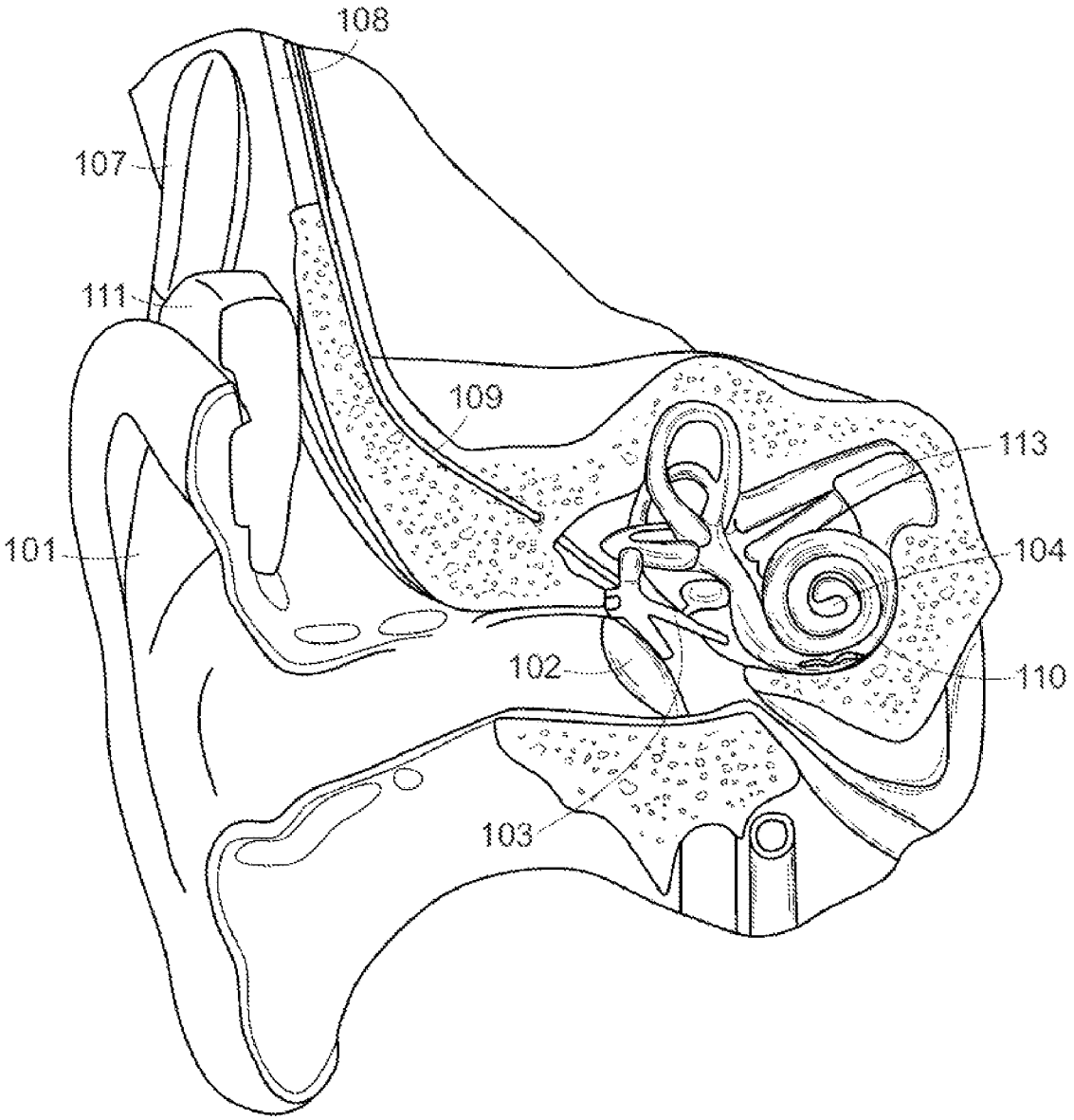

FIG. 1

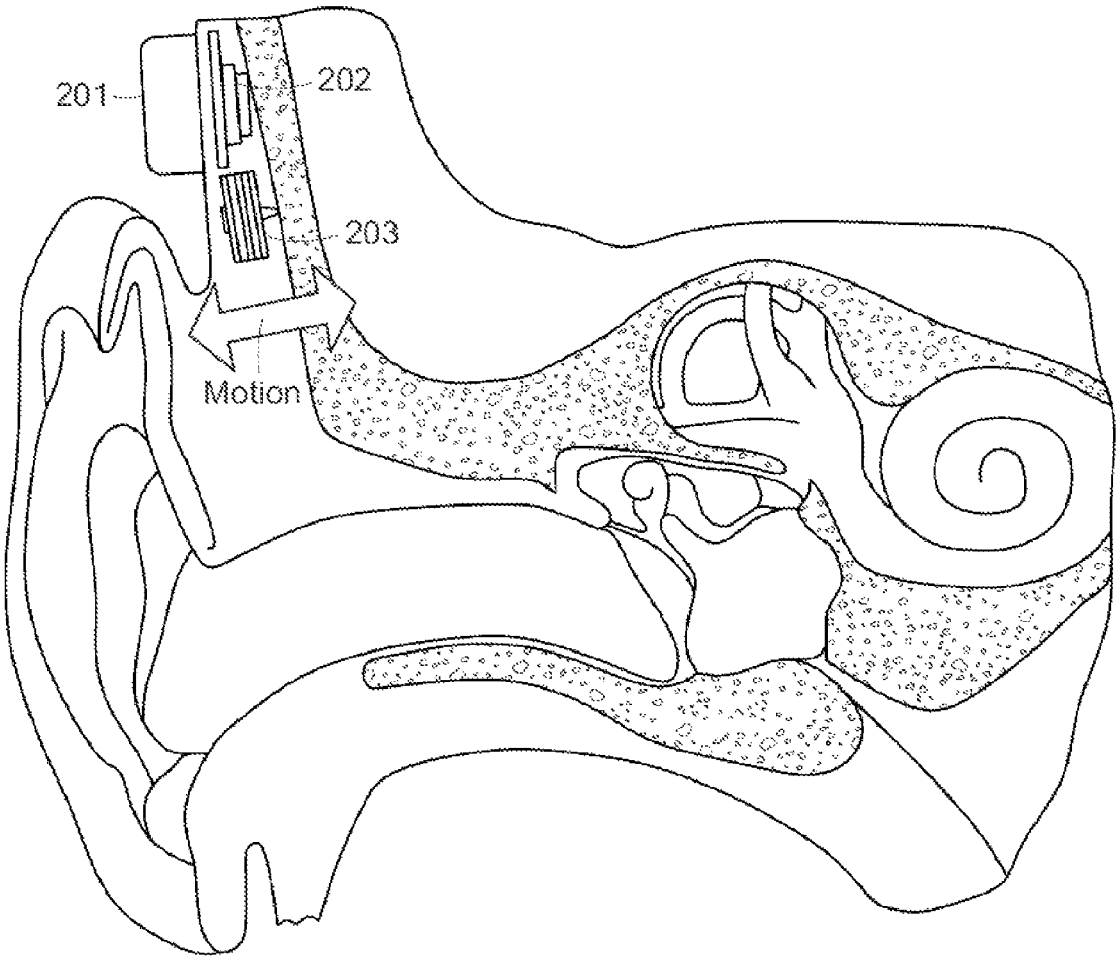

FIG. 2

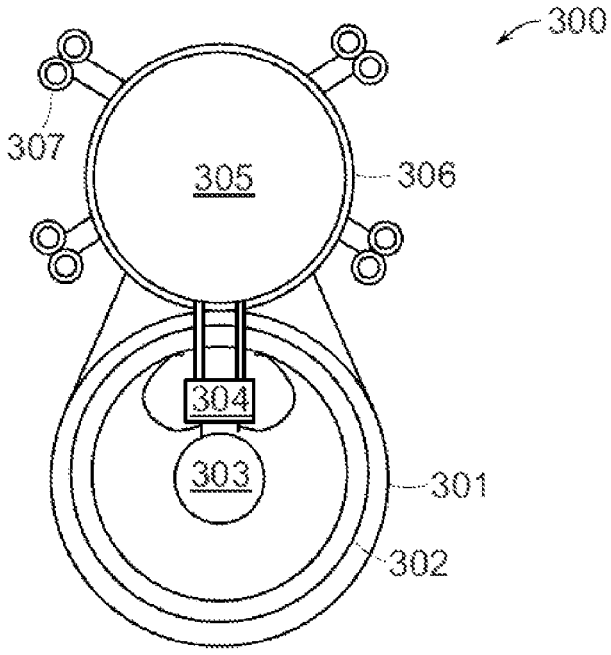

FIG. 3A

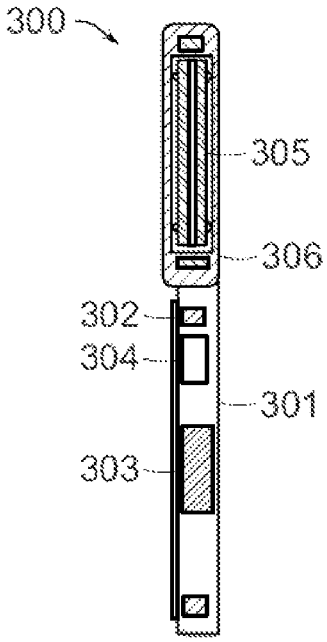

FIG. 3B

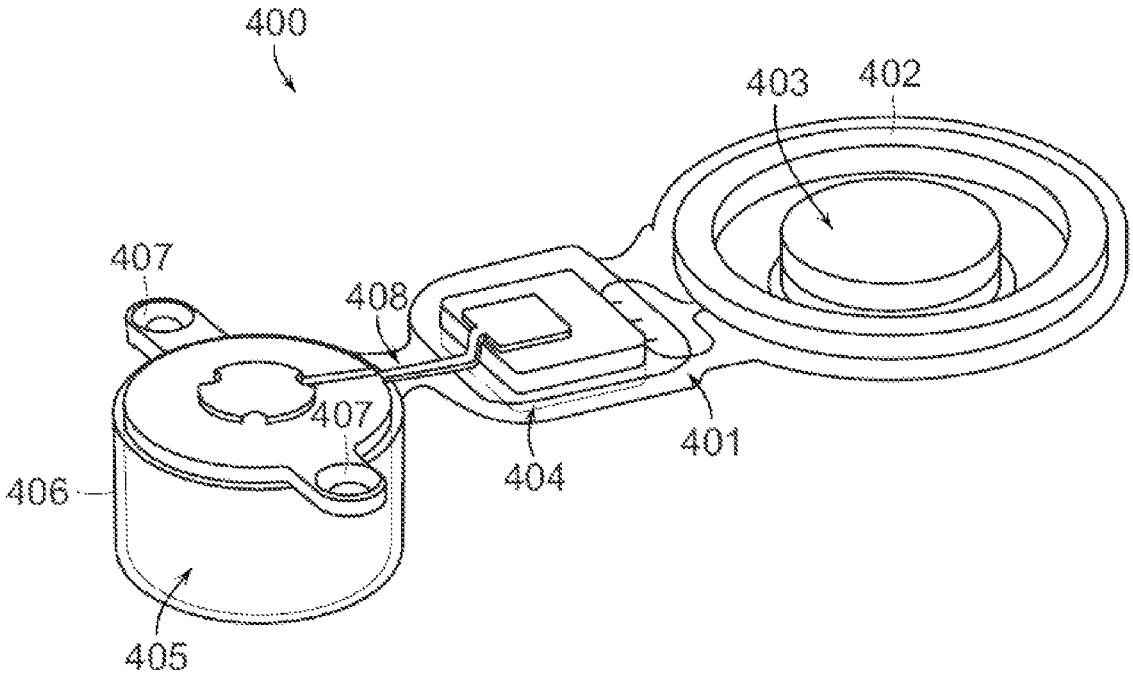

FIG. 4

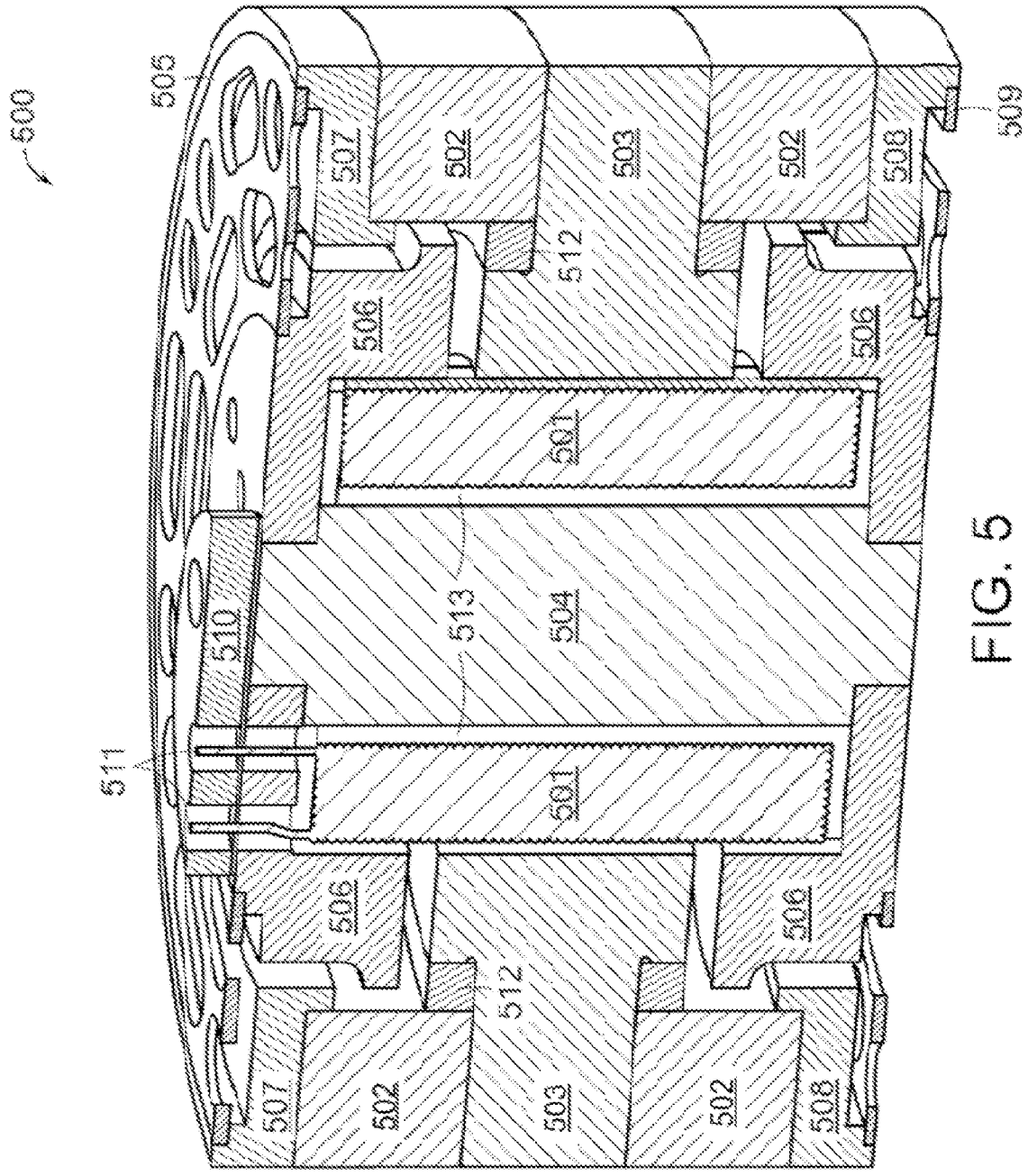
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FIG. 5

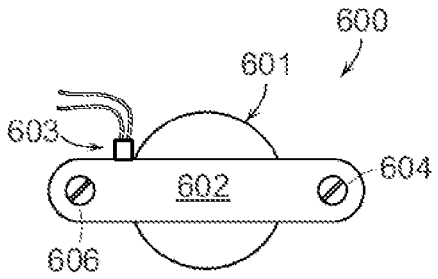

FIG. 6A

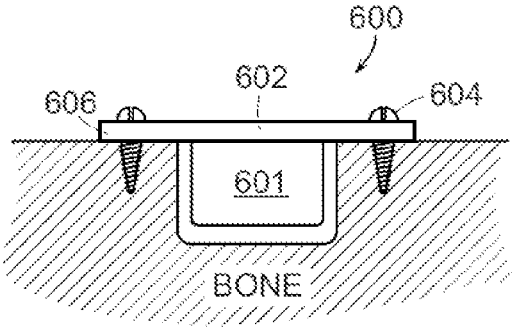

FIG. 6B

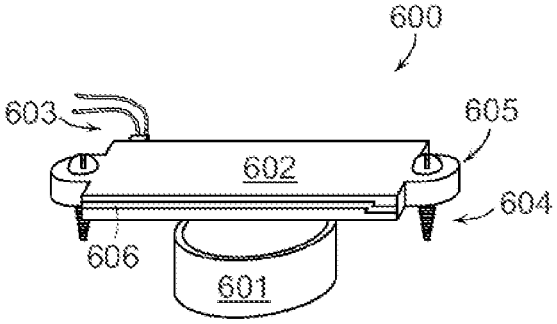

FIG. 6C

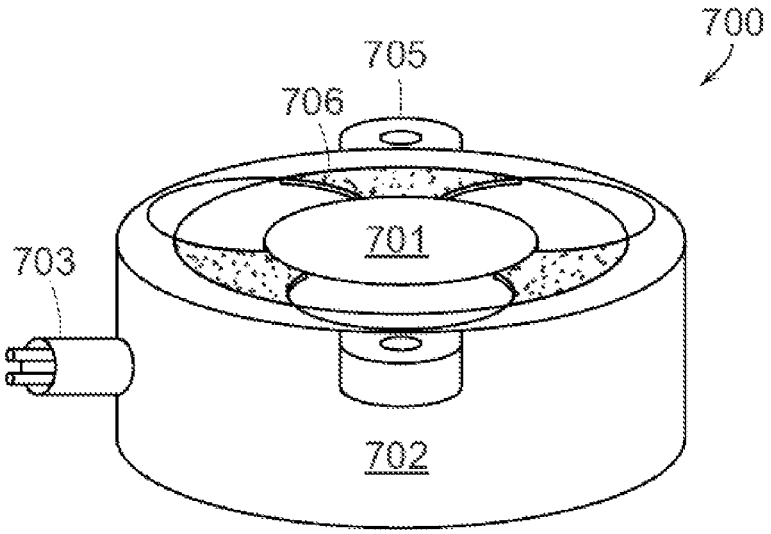

FIG. 7A

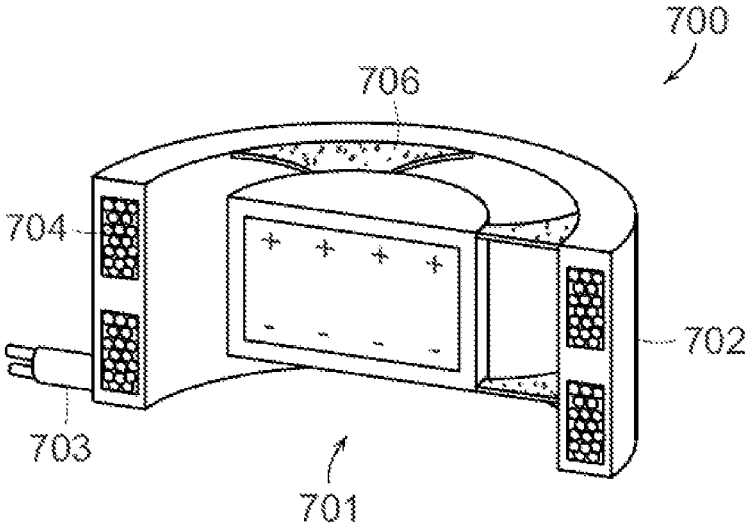

FIG. 7B

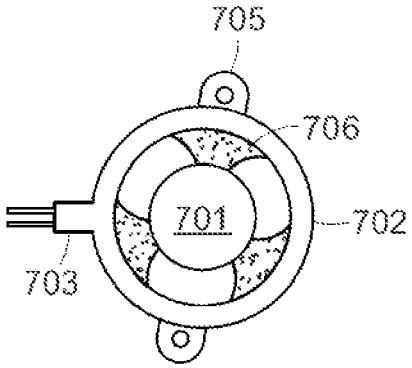

FIG. 7C

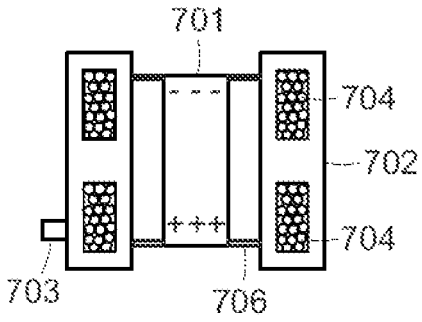

FIG. 7D

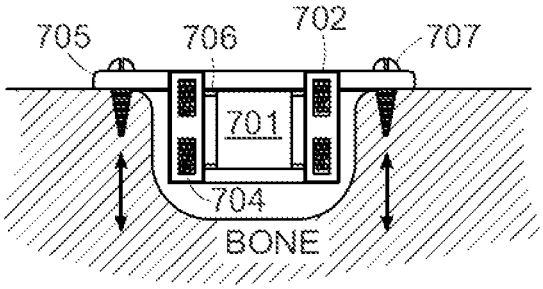

FIG. 7E

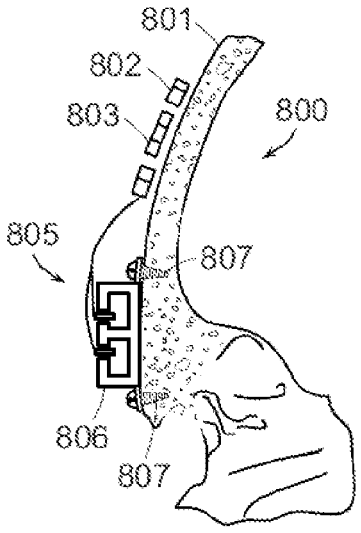

FIG. 8A

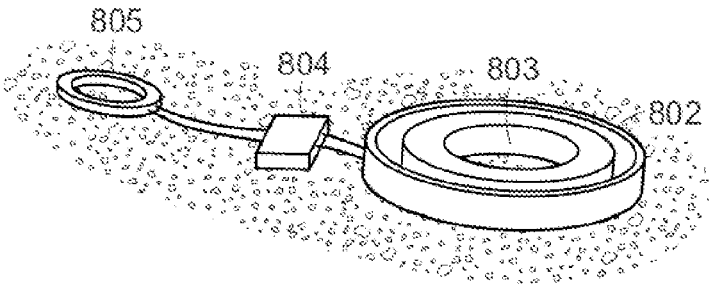

FIG. 8B

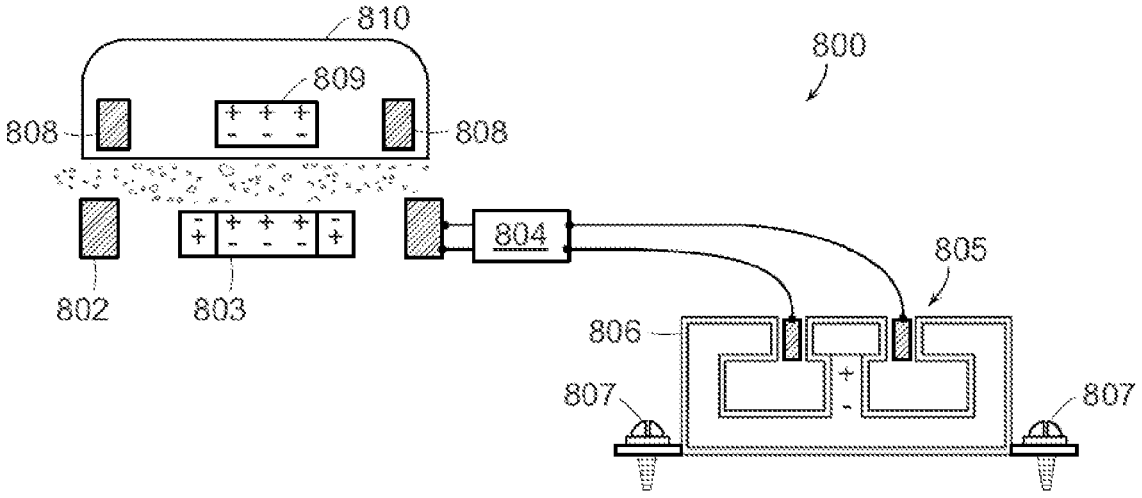

FIG. 8C

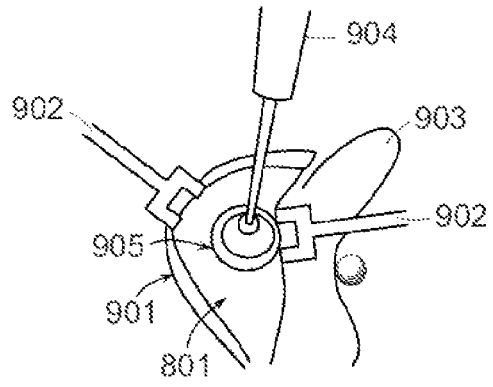

FIG. 9A

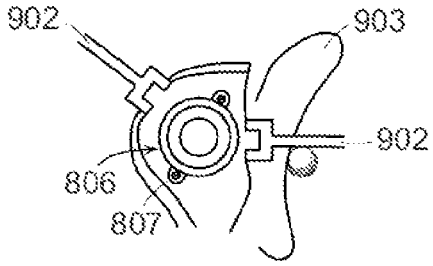

FIG. 9B

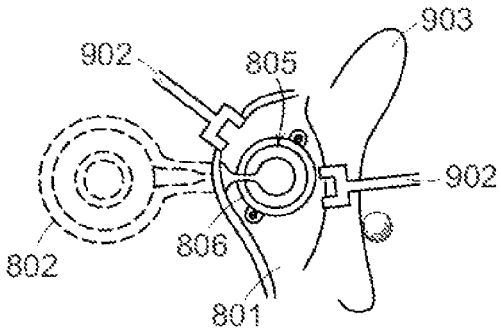

FIG. 9C

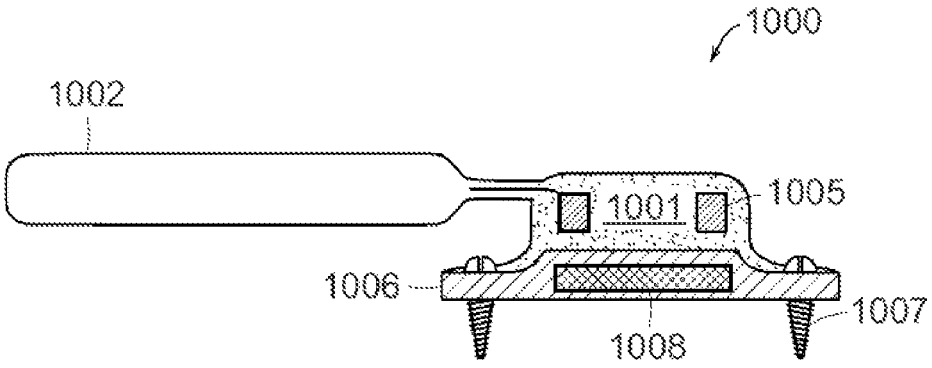

FIG. 10A

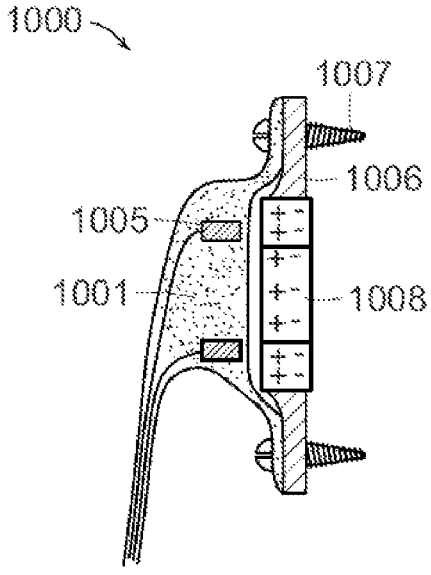

FIG. 10B

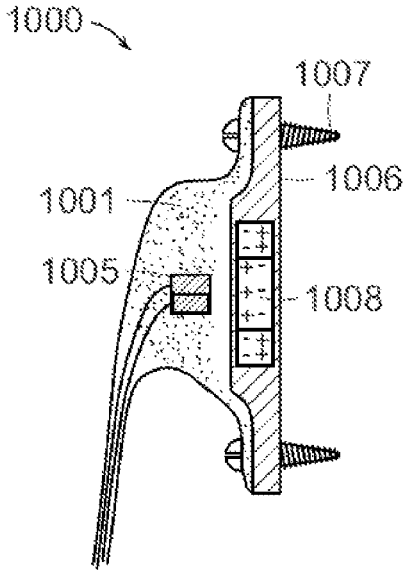

FIG. 10C

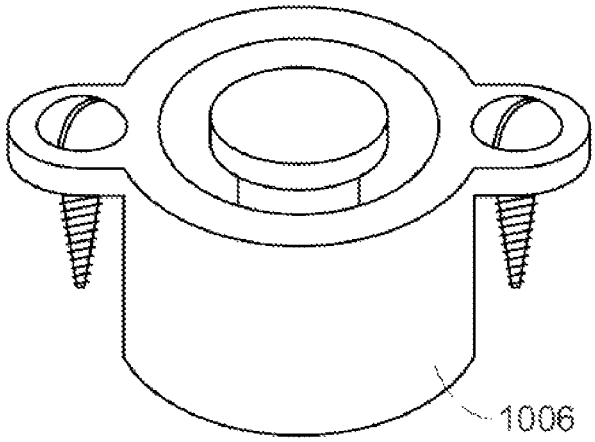

FIG. 11A

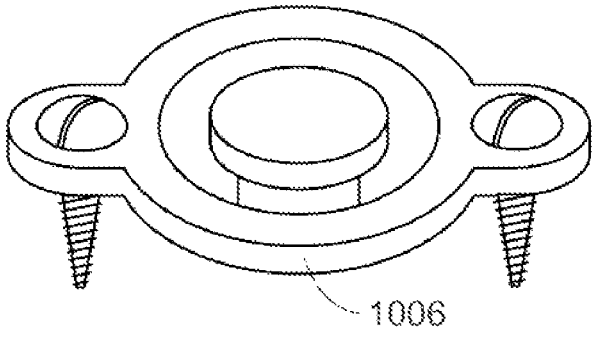

FIG. 11B

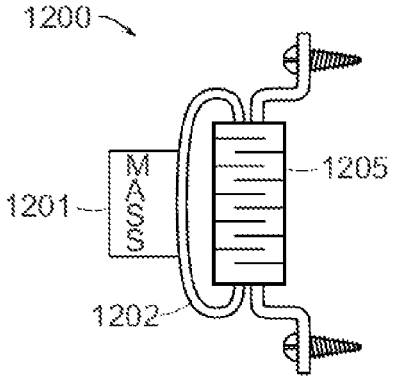

FIG. 12A

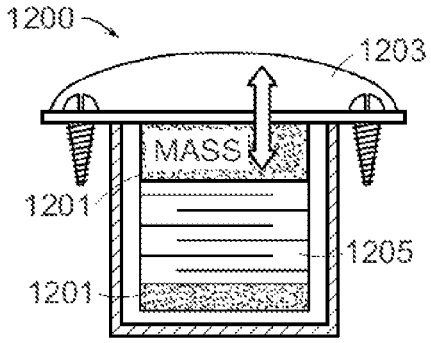

FIG. 12B

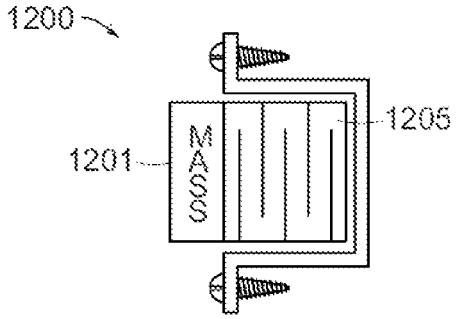

FIG. 12C

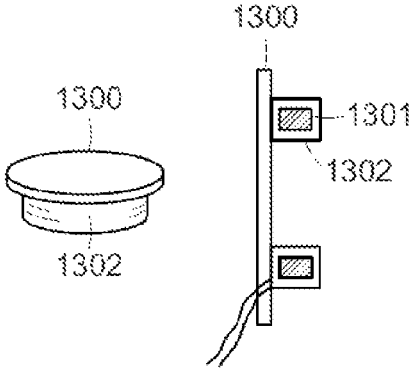

FIG. 13A

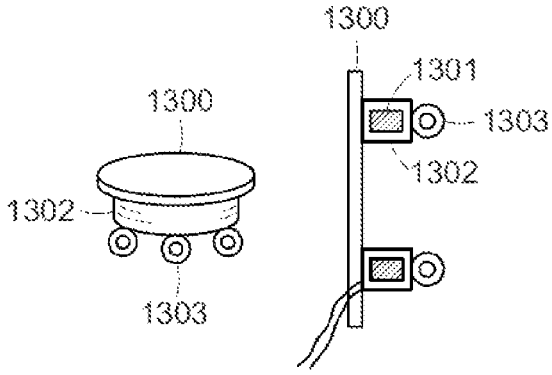

FIG. 13B

MRI SAFE ACTUATOR FOR IMPLANTABLE FLOATING MASS TRANSDUCER

This application is a divisional of U.S. patent application Ser. No. 12/634,940, filed Dec. 10, 2009, which in turn claims priority from U.S. Provisional Patent Application 61/263,150, filed Nov. 20, 2009, and from U.S. Provisional Patent Application 61/227,603, filed Jul. 22, 2009, and from U.S. Provisional Patent Application 61/121,399, filed Dec. 10, 2008, all of which are incorporated herein by reference.

FIELD OF THE INVENTION

The present invention relates to medical implants, and more specifically to a novel bone conduction transducer for an implantable hearing prosthesis.

BACKGROUND ART

A normal ear transmits sounds as shown in FIG. **1** through the outer ear **101** to the tympanic membrane (eardrum) **102**, which moves the ossicles of the middle ear **103** (malleus, incus, and stapes) that vibrate the oval window and round window openings of the cochlea **104**. The cochlea **104** is a long narrow organ wound spirally about its axis for approximately two and a half turns. It includes an upper channel known as the scala vestibuli and a lower channel known as the scala tympani, which are connected by the cochlear duct. The cochlea **104** forms an upright spiraling cone with a center called the modiolar where the spiral ganglion cells of the acoustic nerve **113** reside. In response to received sounds transmitted by the middle ear **103**, the fluid-filled cochlea **104** functions as a transducer to generate electric pulses which are transmitted to the cochlear nerve **113**, and ultimately to the brain.

Hearing is impaired when there are problems in the ability to transduce external sounds into meaningful action potentials along the neural substrate of the cochlea **104**. To improve impaired hearing, various types of hearing prostheses have been developed. For example, when hearing impairment is associated with the cochlea **104**, a cochlear implant with an implanted stimulation electrode can electrically stimulate auditory nerve tissue within the cochlea **104** with small currents delivered by multiple electrode contacts distributed along the electrode. FIG. **1** also shows some components of a typical cochlear implant system which includes an external microphone that provides audio information to an external signal processor **111** where various signal processing schemes can be implemented. The processed data communications signal with the audio information is then converted into a digital data format, such as a sequence of data frames, for transcutaneous transmission by an external transmitting coil **107** to a corresponding receiving coil in an implant processor **108**. Besides extracting the audio information from the data communications signal, the implant processor **108** also performs additional signal processing such as error correction, pulse formation, etc., and produces a stimulation pattern (based on the extracted audio information) that is sent through an electrode lead **109** to an implanted electrode array **110**. Typically, this electrode array **110** includes multiple electrodes on its surface that provide selective stimulation of the cochlea **104**.

When hearing impairment is related to operation of the middle ear **103**, a conventional hearing aid may be used to provide acoustic-mechanical vibration to the auditory system. With conventional hearing aids, a microphone detects sound which is amplified and transmitted in the form of

acoustical energy by a speaker or another type of transducer into the middle ear **103** by way of the tympanic membrane **102**. Interaction between the microphone and the speaker can sometimes cause an annoying and painful a high-pitched feedback whistle. The amplified sound produced by conventional hearing aids also normally includes a significant amount of distortion.

Efforts have been made to eliminate the feedback and distortion problems using middle ear implants that employ electromagnetic transducers. A coil winding is held stationary by attachment to a non-vibrating structure within the middle ear **103** and microphone signal current is delivered to the coil winding to generate an electromagnetic field. A magnet is attached to an ossicle within the middle ear **103** so that the magnetic field of the magnet interacts with the magnetic field of the coil. The magnet vibrates in response to the interaction of the magnetic fields, causing vibration of the bones of the middle ear **103**. See U.S. Pat. No. 6,190,305, which is incorporated herein by reference.

Middle ear implants using electromagnetic transducers can present some problems. Many are installed using complex surgical procedures which present the usual risks associated with major surgery and which also require disarticulating (disconnecting) one or more of the bones of the middle ear **103**. Disarticulation deprives the patient of any residual hearing he or she may have had prior to surgery, placing the patient in a worsened position if the implanted device is later found to be ineffective in improving the patient's hearing.

U.S. Patent Publication 20070191673 and U. S. Provisional Patent Application 61/121,399, filed Dec. 10, 2008, which are incorporated herein by reference, describe driving a relatively large inertial mass to vibrate the skull bone of a hearing impaired patient. As shown in FIG. **2**, a floating mass transducer (FMT) **203** is mechanically connected to the temporal bone of the patient. The mass of the floating mass transducer (FMT) **203** vibrates in response to the audio information in a data communications signal originating from an external processor **201** and transmitted to an implanted receiving coil **202**. Bone conduction of the FMT vibrations through the temporal bone are transduced into fluid motion within the cochlea and perceived as sound.

SUMMARY OF THE INVENTION

Embodiments of the present invention include an implantable hearing prosthesis for a recipient patient. An implantable signal processor is in communication with the receiving coil and converts the communication signal into an electrical stimulation signal. An implantable signal transducer is in communication with the signal processor and includes one or more electromagnetic drive coils for receiving the electrical stimulation signal and a cylindrical transducer magnet arrangement including an inner disk magnet having a first magnetic field direction, and an outer annular magnet surrounding the inner rod magnet and having a second magnetic field direction opposite to the first magnetic field direction. Current flow through the one or more electromagnetic drive coils from the electrical stimulation signal creates a coil magnetic field that interacts with the magnetic fields of the transducer magnet arrangement to create vibration in the transducer magnet which is developed by the signal transducer as a mechanical stimulation signal for audio perception by the patient.

In some embodiments, the signal transducer may include a hermetically sealed transducer housing, which may be sealed by a silicone elastomer and/or may be made of titanium. And the prosthesis may be a middle ear implant device.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. **1** shows structures of a typical ear which includes a cochlear implant.

FIG. **2** illustrates the operating principle of a bone conduction prosthesis.

FIG. **3** shows an example of a prior art bone conduction prosthesis.

FIG. **4** shows an example of an implantable hearing prosthesis according to an embodiment of the present invention.

FIG. **5** shows various structural details of a transducer according to one embodiment of the present invention.

FIG. **6**A-C shows various views of a bone conducting transducer according to one specific embodiment of the present invention based on a piezoelectric inertial mass arrangement.

FIG. **7** shows A-E shows various views of a bone conducting transducer according to one specific embodiment of the present invention based on an arrangement of one or more electromagnetic coils that interact with a permanent magnet inertial mass.

FIG. **8**A-C shows various details of an embodiment having an easily insertable and removable drive transducer.

FIG. **9**A-C shows details of a surgical procedure for inserting an embodiment such as the one shown in FIG. **8**.

FIG. **10**A-C shows various alternative structural details according to specific embodiments.

FIG. **11**A-B shows different height transducer housings according to different embodiments.

FIG. **12**A-C shows structural details of embodiments based on piezoelectric elements.

FIG. **13**A-B shows various structural details of en electromagnetic drive coil according to an embodiment.

DETAILED DESCRIPTION OF SPECIFIC EMBODIMENTS

FIG. **3** shows elements of an implantable hearing prosthesis as described, for example, in U.S. Patent Publication 20070191673 ("Ball '673"), which is based on driving a relatively large mass to vibrate the skull bone of a hearing impaired patient. Bone conduction of these vibrations is transduced into fluidic vibration within the cochlea that is sensed by the patient as sound. More specifically, FIG. **3**A shows a top plan view and FIG. **3**B shows a side cross-section view of an implantable hearing prosthesis **300** using an inertial mass-based bone conduction transducer. A silicone elastomer receiver housing **301** contains a receiving coil **302** that transcutaneously receives communications signals from the external audio processor, and a holding magnet **303** that cooperates with a corresponding external magnet to hold the external audio processor in correct position over the receiving coil **302**. An implant signal processor **304** receives the communications signals from the receiving coil **302** and produces a corresponding electrical stimulation signal to a bone conduction transducer **305**, specifically, a dual opposing magnet type floating mass transducer (FMT), which is enclosed in a titanium transducer housing **306**. Mounting of the transducer housing **306** to the skull bone is accomplished by multiple pairs of attachment ears **307** which are surgically mounted to the bone with connecting screws. The FMT mass of the bone conduction transducer **305** vibrates in response to the electrical stimulation signal from the implant signal processor **304**, which in turn causes inertial vibration of the transducer housing **306**. The housing vibrations are transduced through the temporal bone by bone conduction into fluid motion within the cochlea and perceived as sound.

While an improvement in the field, the implantable hearing prosthesis **300** of Ball '673 is not without issues. For example, the Ball '673 implantable hearing prosthesis **300** has multiple mounting holes which require a high degree of planarity in the bone surrounding the implantation site. And the Ball '673 implantable hearing prosthesis **300** is configured such that in a relaxed state, the receiver housing **301** and the transducer housing **306** are biased to lie in a single plane. Thus, when implanted onto the curved skull bone of a recipient patient, this existing bias exerts a force that tends to pull the two housings back into a common plane, away from the curvature of the underlying skull bone.

Embodiments of the present invention are directed to an implantable bone conduction hearing prosthesis with various improvements over the earlier Ball '673 device. FIG. **4** shows one example of such an implantable hearing prosthesis **400** having a silicone elastomer receiver housing **401** (e.g., about 4.5 mm thick) that contains a receiving coil **402** and a holding magnet **403**. Implant signal processor **404** receives the communications signals from the receiving coil **402** and produces a corresponding electrical stimulation signal to a bone conduction transducer **405**, which is a dual opposing magnet type floating mass transducer (FMT). The FMT mass of the bone conduction transducer **405** is enclosed in a titanium transducer housing **406**, which typically may be about 17 mm across and about 11 mm in depth.

FIG. **5** shows various internal structural details of a bone conduction transducer **500** for an implantable hearing prosthesis **400** as shown in FIG. **4**. An axially central electromagnetic coil **501** is surrounded by a coil spacer **513**, a central base core **504**, and core spacer **506**. The central base core **504** and core spacer **506** are made of soft iron that increases the magnetic coupling of the magnetic field to provide a magnetic conduction path for the coil flux. Radially surrounding central core subassembly is a moveable subassembly of one or more ring-shaped permanent magnets **502** assembled together with a soft iron magnet carrier **503** and one or more magnet spacers **512**. This moveable subassembly is attached to a top suspension subassembly of a top membrane spring **505** together with a soft iron top lid **507**, and a bottom suspension subassembly of a bottom membrane spring **509** together with a soft iron bottom lid **508**. The bias point of the permanent magnets **502** can be kept in a safe range (high B-field, low H-field) with respect to demagnetization from aging or external magnetic fields.

Operation of the transducer **500** is based on employing a motion constraint (e.g., the self-centering parallel membrane springs **505** and **509**) to create a linear-mode inertial drive of electrical stimulation signals. The electrical stimulation signal from the implant signal processor **404** is received by coil feeds **511** in a coil feed clip **510** and developed by the electromagnetic coil **501** and base core **504**. This produces a coil magnetic field that interacts with the base core **504**, the one or more permanent magnets **502**, and magnet carrier **503**. The one or more permanent magnets **502** and magnet carrier **503** vibrate in response to the stimulation signal. This vibration of the transducer **500** is then coupled to the adjacent bone for bone conduction to the cochlea.

In addition, the arrangement of structural features in the transducer **500** avoids magnetic short circuits due to the air gaps between the moveable permanent magnets **502** and the non-moveable electromagnetic coil **501** and core spacer **506**.

The non-magnetic membrane springs **505** and **509** prevent these air gaps from collapsing when the transducer **500** is excited by an electrical stimulation signal (one of the moveable parts would magnetically stick to one of the core parts). Instead, when there is no stimulation signal, the forces in the air gaps generated by the magnetic bias flux compensate and balance each other. When an electrical stimulation signal is present and providing excitation to the transducer **500**, the flux density will weakened in one of the air gaps and boosted in the other. The resulting net force is non-zero and the moveable subassembly moves in response. Vice versa, the transducer **500** can be used to generate a corresponding electrical signal from vibrational excitation, for example, to act as an implant sensor or to generate energy for the implant system. Closed-loop control applications may be realized by fitting the transducer **500** with a sensing element.

Inductance can be minimized in the electromagnetic coil **501** by controlling stray magnetic flux. Mechanical resonance frequency of the transducer **500** also can be fine-tuned in various ways such as by spring trimming with a cutting laser. Eddy currents can be used in the transducer **500** to provide dampening of resonance peaks by magnetically non-conductive short circuit elements. Some embodiments may also immerse components in a viscous fluid for additional dampening.

Compared to prior inertial transducers, the transducer **500** in FIG. **5** better maximizes the inertia of the involved masses (and also thereby achieving lower resonance frequencies) by having the moveable subassembly of the permanent magnets **502** and magnet carrier **503** radially outside the electromagnetic coil **501** and central base core **504**. Similarly, having loss-generating components such as the electromagnetic coil **501** closer to the axial center of the transducer **500**, higher efficiency is enjoyed as compared to prior art arrangements.

Such an arrangement is also easily manufacturable because of the rotationally symmetric design, use of relatively massive non-laminated yoke components with low electrical conductivity. In addition, it may be useful to use multiple separate yoke parts and/or use components with self-centering characteristics. Radial slots in one or more of the yoke components may also be useful for minimizing the influence of eddy currents. Such an arrangement also minimizes distortion compared to prior art designs by intentionally introducing ferromagnetic saturation in certain yoke regions by stabilizing constant bias flux. Besides use for bone conduction hearing applications, a transducer **500** may be useful in other types of applications such as for bone healing, a membrane pump, energy harvesting, active vibration dampening, hydraulic valves, loudspeakers, and/or vibration exciter.

Returning to FIG. **4**, the receiver housing **401** and the transducer housing **406** are connected at an unbiased pivot point **408**. The unbiased pivot point **408** allows the receiver housing **401** to be bent out of the plane containing the upper surface of the transducer housing **406** so that it lies correctly in a relaxed condition in proper position under the skin, without the kind of undesirable bias force found in the devices described in Ball '863 that tends to flex the receiver housing back towards the plane of the transducer housing. Such unbiased bending of the housings relative to each other is helpful for accommodating different sizes of patient skulls and corresponding varying amounts of skull bone curvature. Some skulls are relatively smaller and therefore need relatively more bend between the housings, while other skulls are relatively larger and little or no bending of the housings may be needed. In one specific embodiment, the receiver housing **401** can be bent without residual biasing force up to 180 degrees from a 90 degree superior to a 90 degree inferior position in relation to the transducer housing **406**.

Mounting of the transducer housing **406** to the skull bone is accomplished by two single mounting points **407** which are opposite to each other on the outer perimeter of the transducer housing **406** so as to couple the mechanical vibration signal from the bone conduction transducer **405** via bone conduction to the cochlea. The use of two single mounting points **407** in the implantable hearing prosthesis **400** avoids some of the bone planarity issues associated with the multiple mounting point embodiments described in Ball '673. The mounting points **407** may be secured to the skull bone with single-use self-tapping bone screws, e.g., 6-8 mm in length. Use of self-drilling screws may cause micro-fractures in the bone. In some patients, it may be preferred to use different length bone screws in each mounting point **407**.

An implantable hearing prosthesis **400** can be implanted in a relatively simple surgical procedure that may take as little as 30 minutes. The surgeon creates a skin incision over the desired location of the device, a bone bed is prepared, and screw holes are pre-drilled for the mounting screws. An implant template may be useful for these steps to aid in preparation of the proper size and shape bed and/or to act as a drill guide for drilling of the screw holes. The hearing prosthesis **400** is inserted into position and secured with the mounting screws which are tightened to a defined torque. Then the receiving housing **401** is bent into proper position at the unbiased pivot point **408**, and the incision is closed.

FIG. **6**A-C shows various views of one specific embodiment of a bone conduction transducer **600** for an implantable hearing prosthesis which uses one or more piezoelectric members **606**. Signal input **603** is a feed-through wiring arrangement that receives an electrical stimulation signal from an implant signal processor. A transducer housing **601** is suspended below the piezoelectric members **606** in a prepared bone recess which surrounds the inertial mass housing **601**. The piezoelectric members **606** respond to the electrical stimulation signal with corresponding mechanical vibrations. The mechanical vibrations are also imparted to the transducer housing **601** that is suspended below the piezoelectric members **606** and in effect amplifies the magnitude of the mechanical vibrations. The mechanical vibrations of the transducer housing **601** and the piezoelectric members **606** are coupled through mounting points **606** and corresponding connecting screws **604** which attach to the skull bone (such as the cortical bone or the temporal bone of the patient), and carried by bone conduction to the cochlea to be perceived as sound.

FIG. **7**A-E shows various views of another embodiment a bone conduction transducer **700** of an implantable hearing prosthesis based on an inertial mass housing arrangement which includes one or more electromagnetic coils **704** surrounding a permanent magnet **701** for responding to the electrical stimulation signal with the corresponding mechanical vibrations. In this case, the electromagnetic coils **704** are contained in a hermetic cylindrical coil housing **702** made of titanium within which is the inertial mass of the permanent magnet **701**. The permanent magnet **701** is flexibly suspended within the center of the coil housing **702** by a flexible connector member **706**. In the example shown, the flexible connector member **706** is in the specific form of arcuate segments of a flexible diaphragm.

Operation of this embodiment can most clearly be seen from the view shown in FIG. **6**E. The electromagnetic coils **704** respond to the electrical stimulation signal with a varying electromagnetic field that in turn interacts with the permanent magnet **701** to generate corresponding mechanical vibration

that moves the permanent magnet **701** up and down. The mechanical vibrations are coupled through the flexible connector member **706** to the coil housing **702** to the mounting points **705** and corresponding connecting screws **707** which attach to the skull bone (such as the cortical bone or the temporal bone of the patient). The skull bone then conducts the audio information of the mechanical vibrations to the cochlea.

FIG. **8**A-C shows various views of another embodiment of the present invention. An external processor **810** contains one or more sensing microphones for sensing the acoustic environment around a patient user and generating a corresponding microphone signal. From the microphone signal the external processor generates a representative communication data signal which is transcutaneously transmitted by an external transmitting coil **808** to an implanted receiving coil **802**. An implant magnet **803** within the receiving coil **802** magnetically interacts with a corresponding external holding magnet **809** within the transmitting coil **808** to hold the external processor **810** in a correct position. An implantable signal processor **804** converts the communication data signal from the receiving coil **802** into a representative electrical stimulation signal. An implantable transducer housing **806** is fixedly attachable to the skull bone **801** of the patient. An implantable drive transducer **805**, in this case an electromagnetic drive coil, is in communication with the signal processor **804** and removably engageable with the transducer housing **806** for applying to the transducer housing **806** a mechanical vibration signal based on the electrical stimulation signal for audio perception by the patient.

In the embodiment shown in FIG. **8**, transducer housing **806** is fixedly attached to the skull bone **801** during a surgical procedure such as the one shown in FIG. **9**A-C. In FIG. **9**A, a surgical incision **901** is made in the patient's skin around the site of the transducer housing **806** behind the ear auricle **903**. Retractors **902** pull back the skin and ear auricle **903** from the surgical site to provide access for a surgical drill **904** to prepare a recessed bone well in the skull bone **801**. The transducer housing **806** is then fixed in place in the bone wells by a pair of radially opposed bone screws **807**, after which the remainder of the prosthetic system is implanted including inserting the drive transducer **805** into the ready transducer housing **806**. Then later, if any portion of the system needs replacement, the drive transducer **805** can be easily withdrawn from the transducer housing **806** during a simple surgical procedure without disturbing the existing connection with the patient skull bone **801**.

FIG. **10**A-C shows an embodiment of an implantable prosthesis system **1000** wherein a silicone elastomer mold **1001** encases an electromagnetic drive coil **1005** (e.g., made polyimed coated gold wire) together in a sealed engagement with a low-profile transducer housing **1006**. The silicone elastomer mold **1001** provides protective encasing of the drive coil **1005** and may also act as a spring to enhance long term stability and reduce signal distortion. The low-profile transducer housing **1006** includes a drive magnet **1008** which interacts with the electromagnetic drive coil **1005** to couple the mechanical vibration signal to the underlying skull bone. FIG. **10**C shows a variation in which the drive magnet **1008** has a coaxial double magnet arrangement where the center has a first magnetic polarity and the outer ring has a second opposite magnetic polarity. In this embodiment, the drive coil **1005** may be arranged correspondingly, for example, in a tight central structure that interacts mainly with the center of the drive magnet **1008**.

FIG. **11**A-B shows embodiments having different height profiles on the transducer housing **1106**. In both embodiments, the transducer housing **1106** forms a hermetically sealed can, but in the embodiment shown in FIG. **11**A, the transducer housing is much higher, e.g., about the same as the diameter of the housing, typically around 10 mm. FIG. **11**B shows a lower height transducer housing **1106** which has a height much less than the diameter of the housing, e.g., about 5 mm. Where the height of the transducer housing **1106** is higher such as shown in FIG. **11**A, it is more likely that a recessed bone well may be needed where the housing is fixed the skull bone in order to accommodate the relatively high profile of the housing. On the other hand, where the height of the transducer housing **1106** is lower as shown in FIG. **11**B, it may be that the housing can be correctly attached to the skull bone with needing a recessed bone well, thereby making surgical installation much easier.

In some embodiments, the drive transducer may be a piezoelectric transducer. For example, FIG. **12**A shows an embodiment of a drive transducer **1200** having an inertial mass **1201** that is coupled to a piezoelectric stack **1205** containing piezoelectric elements stacked parallel to the surface of the skull bone. In this embodiment, a coupling bow **1202** of stiff material (e.g., titanium) provides the mechanical connection of the inertial mass **1201** to the piezoelectric stack **1205**.

FIG. **12**B shows an embodiment where the drive transducer **1200** includes opposing inertial masses **1201** at either end of a piezoelectric stack **1205** containing piezoelectric elements stacked perpendicular to the surface of the skull bone. A coupling diaphragm **1203** of stiff material (e.g., titanium) mechanically connects the drive transducer **1200** to the skull bone. FIG. **12**C shows another embodiment where the drive transducer **1200** includes a single inertial mass **1201** at one end of a piezoelectric stack **1205** containing piezoelectric elements stacked perpendicular to the surface of the skull bone.

In some embodiments, shown for example in FIG. **13**A-B, the drive coil **1301** may be covered by an encapsulation layer **1302** of biocompatible material such as silicone or acrylic. In the specific embodiments shown in FIG. **13**A-B, the outer axial end of the drive coil **1301** has a sealing lens **1300** of biocompatible material which helps with the installation of the drive coil **1301** in the transducer housing. Such a sealing lens **1300** may also act as a spring to help minimize signal distortion. The sealing lens **1300** in FIG. **13**B also includes a separate coupling spring **1303** incorporated into the encapsulation layer **1302** at the inner axial end of the drive coil **1302** for coupling the drive coil **1302** to the transducer housing with minimal distortion and long term durability. In other embodiments, the transducer housing may include such a coupling spring.

Embodiments of the present invention may be most appropriate for patients with conductive hearing impairment exhibiting mixed hearing loss with bone conduction thresholds better than or equal to 45 dB HL at various audiogram evaluation frequencies. A physician considering use of such a device should fully assess the potential risks and potential benefits for the patient, bearing in mind the patient's complete medical history, and exercising sound medical judgment. Embodiments may be contraindicated for patients with an existing mastoid condition that precludes attachment of the transducer, patients with retrocochlear or central auditory disorders, and/or patients with any known allergies to any of the materials used in the device.

Although various exemplary embodiments of the invention have been disclosed, it should be apparent to those skilled in the art that various changes and modifications can be made which will achieve some of the advantages of the invention without departing from the true scope of the invention.

What is claimed is:

1. An implantable hearing prosthesis for a recipient patient, the prosthesis comprising:

a receiving coil for transcutaneous receiving of an externally generated communication signal;

an implantable signal processor in communication with the receiving coil for converting the communication signal into an electrical stimulation signal;

an implantable inertial mass signal transducer in communication with the signal processor and including:

i. one or more electromagnetic drive coils for receiving the electrical stimulation signal;

ii. a cylindrical transducer magnet arrangement operating as an inertial mass and including an inner disk magnet having a first magnetic field direction, and an outer annular magnet surrounding the inner disk magnet and having a second magnetic field direction opposite to the first magnetic field direction;

wherein the one or more electromagnetic drive coils is arranged in a central structure over the inner disk magnet so that current flow through the one or more electromagnetic drive coils from the electrical stimulation signal creates a coil magnetic field that interacts mainly with the magnetic field of the inner disk magnet so as to create vibration in the transducer magnet arrangement which is developed by the signal transducer in response to the inertial mass of the transducer magnet arrangement as a mechanical stimulation signal for audio perception by the patient.

2. A prosthesis according to claim **1**, wherein the signal transducer includes a hermetically sealed transducer housing.

3. A prosthesis according to claim **2**, wherein the transducer housing is sealed by a silicone elastomer.

4. A prosthesis according to claim **2**, wherein the transducer housing is made of titanium.

5. A prosthesis according to claim **1**, wherein the prosthesis is a middle ear implant device.

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